LEVEL II (12)

LONG-BONE INJURY CRITERIA FOR USE WITH THE ARTICULATED TOTAL BODY MODEL

TIM HIGHT, PhD

JANUARY 1981

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NOTICES

TECHNICAL REVIEW AND APPROVAL

AFAMRL-TR-81-3

This report has been reviewed by the Office of Public Affairs (PA) and is releasable to the National Technical Information Service (NTIS). At NTIS, it will be available to the general public, including foreign nations.

This technical report has been reviewed and is approved for publication.

FOR THE COMMANDER

HENNING E. VON GIERKE , Dr. Eng.
Director
Biodynamics and Bioengineering Division
Air Force Aerospace Medical Research Laboratory

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An existing computer program, the Articulated Total Body (ATB) model, calculates joint loadings, contact forces and segment positions for a 15-segment model of a human body subjected to high accelerations and wind loads. This project has attempted to augment this model by providing a means of predicting fracture in the long bones. Failure criteria which have been suggested in the literature are discussed, and a new set of criteria is developed. These new [continued on reverse]

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20. (Continued)

fracture criteria are based on reported stress vs. strain and strain rate properties for human bone in compression, and have been extended to also provide tension and shear loading criteria. Two forms of the criteria for ultimate stress are developed, one in terms of stress rate and the other in terms of pulse length. Demonstration results are presented for an aircraft ejection simulation which shows the maximum stresses and allowable stresses (both for stress rate and pulse length) as functions of time for the left lower arm.

PREFACE

The work described in this report was accomplished by Dr. Tim Hight
of Duke University, Durham, NC, under the Air Force Systems Command,
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INTRODUCTION

The simulation of complex, dynamic interactions through the use of computer solutions of mathematical models has become an extremely useful and cost-effective research and development tool. One of the most intriguing areas of application is in the simulation of the reaction of the human body to harsh environments. These simulations attempt to recreate or predict the forces and motions experienced by a body in a high-acceleration event such as an automobile impact. The Air Force has particular interest in the reaction of the subjects to emergency ejections from high-speed aircraft. This is an extremely hazardous environment with a very high injury rate (1) and is ideally suited for computer simulation.

The Mathematics and Analysis Branch of the Air Force Aerospace Medical Research Laboratory (AFAMRL) has been using a human body simulation program known as the Articulated Total Body (ATB) model for some years now. This model predicts the gross motion of the body segments (15 segments used), considered as rigid bodies connected by non-linear springs, to any acceleration or impact loading. Forces and moments at the joints and contact forces are calculated, as well as linear and angular displacements, velocities and accelerations. The Air Force has been using this model to study the reactions of the body to ejection-seat accelerations and to the high wind velocities encountered just after ejection.

One of the limitations of the ATB model has been the lack of any injury criteria which might be used to judge the severity of loading on the long bones. The high accelerations, wind flail and segment impacts lead to high loadings in the extremities, but it has not been possible to interpret these in terms of any injury potential. Specifically, estimates of the likelihood of bone fracture are desired.

The serious investigation of bone strength has been going on through most of this century, but there is still considerable disagreement over the results. It is fairly clear, however, that bone is a viscoelastic material, and that the strength and other properties are time-dependent.

What is desired, therefore, is to incorporate into the ATB model some mechanism for interpreting the loading on the extremities in terms of a bone fracture criterion which includes a loading rate dependency.

OBJECTIVES

The goal of this Summer Faculty Research Project has been to implement a modification to the current ATB model which would allow a prediction of injury to limb segments based on simplified injury criteria. The subroutines were to be designed to facilitate the extension of the injury criteria to more complex representations. Specifically, this project was to define the severity of loading on a limb, based on joint loadings, limb accelerations, and forces due to impacting bodies.

There were to be two major aspects to this program, 1) the establishment of meaningful injury mechanisms for the long bones; and 2) the assembly of the load data from joints, accelerations and impacting bodies into an appropriate deformable limb model. A definitive injury description was not expected from this study, but the program was to be structured from the start to allow for inclusion of more sophisticated models as they were developed.

INJURY CRITERIA

There are two basic approaches one may take in developing injury criteria for something as generally defined as the "human extremities." These could loosely be thought of as an empirical vs an analytic approach, although these labels indicate more the starting point of an approach rather than an exclusive technique. In the "analytic" approach, one starts with the geometric and material properties of the bones and builds up a model which can interact with an external environment to produce stresses which are compared to some model of ultimate or fracture properties. In the empirical approach, tests are carried out on surrogate specimens (either cadaver or animal) and the loads which produce injury are recorded. The injury-producing loads then define the injury criteria. The automobile industry relies heavily on the second approach, but has so far restricted the range of extremity loading considered to knee impacts.

Because most extremity injuries are not life threatening, there has been, up until recently, little attention paid toward developing specific injury criteria for them. There is now only one such accepted (i.e. federally imposed) standard, and that is the NHTSA knee impact criterion. This is currently set at a maximum of 1700 lbs axial force (irrespective of rate) as measured in the femur load cell of an instrumented dummy (2). This particular criterion came about in response to the relatively high incidence of knee impact injuries and is based on a large number of experimental observations of injury-producing loads.

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In the mid to late 60's, papers began to appear dealing with the problems of knee impacts and leg injuries in automobile accidents. Cooke and Nagel reported on a series of controlled cadaver tests in 1969 (3). In the same publication Wilson reported on sled simulation studies of knee impacts carried out by General Motors (4). Knee impacts and femur loads have received increased attention since that time (5, 6, 7, 8, 9).

In 1973 King, Fan and Vargovick (10) proposed adding a time-dependent feature to the femur injury criterion for loading pulse lengths below 30-50 msec. Their review of the literature indicated a significant increase in breaking load in the femur for pulse lengths below this range. These authors proposed using the time-dependent bone fracture data generated by McElhaney (11) as the basis for a time-dependent criterion. After converting strain rate to an appropriate pulse length ($T = \epsilon / \dot{\epsilon}$ for constant strain rate), McElhaney's relation between maximum stress and strain rate,

$$\sigma_u = 4200 \log \dot{\epsilon} + 33000 \tag{1}$$

was matched to a breaking force of 1650 lb at 50 msec. This leads to the following relationship

$$F = 1370 - 215 \log T \tag{2}$$

where F is femur force in lbs and T is the pulse duration in seconds. Equation 2 predicts a fracture load of 1900 lb for a 3-msec pulse and 2000 lb for a 1-msec pulse.

An alternative approach was taken more recently by Viano (12). This is also based on a survey of data from impact tests. The proposed criterion is a constant 2000 lb load for $T > 20$ msec (2000 lb is the proposed new federal standard), and equation 3 for pulses below that level

$$F = 5200 - 160T \qquad T < 20 \text{ msec} \tag{3}$$

A comparison of these two criteria in Figure 1 shows that the King criterion is much more conservative. The data presented by Viano indicated that his criterion would be below most, but not all, fractures.

The major difficulty with these criteria is their very restrictive definition of loading - namely knee loads producing basically axial loads in the femur. In order to have the most general injury criteria possible, the analytic approach has been chosen. This leads us first to a close examination of material properties.

What is needed is some model which will indicate when a calculable parameter has exceeded its allowable range. In standard strength of materials this would be the maximum normal stress, von Mises-Henky
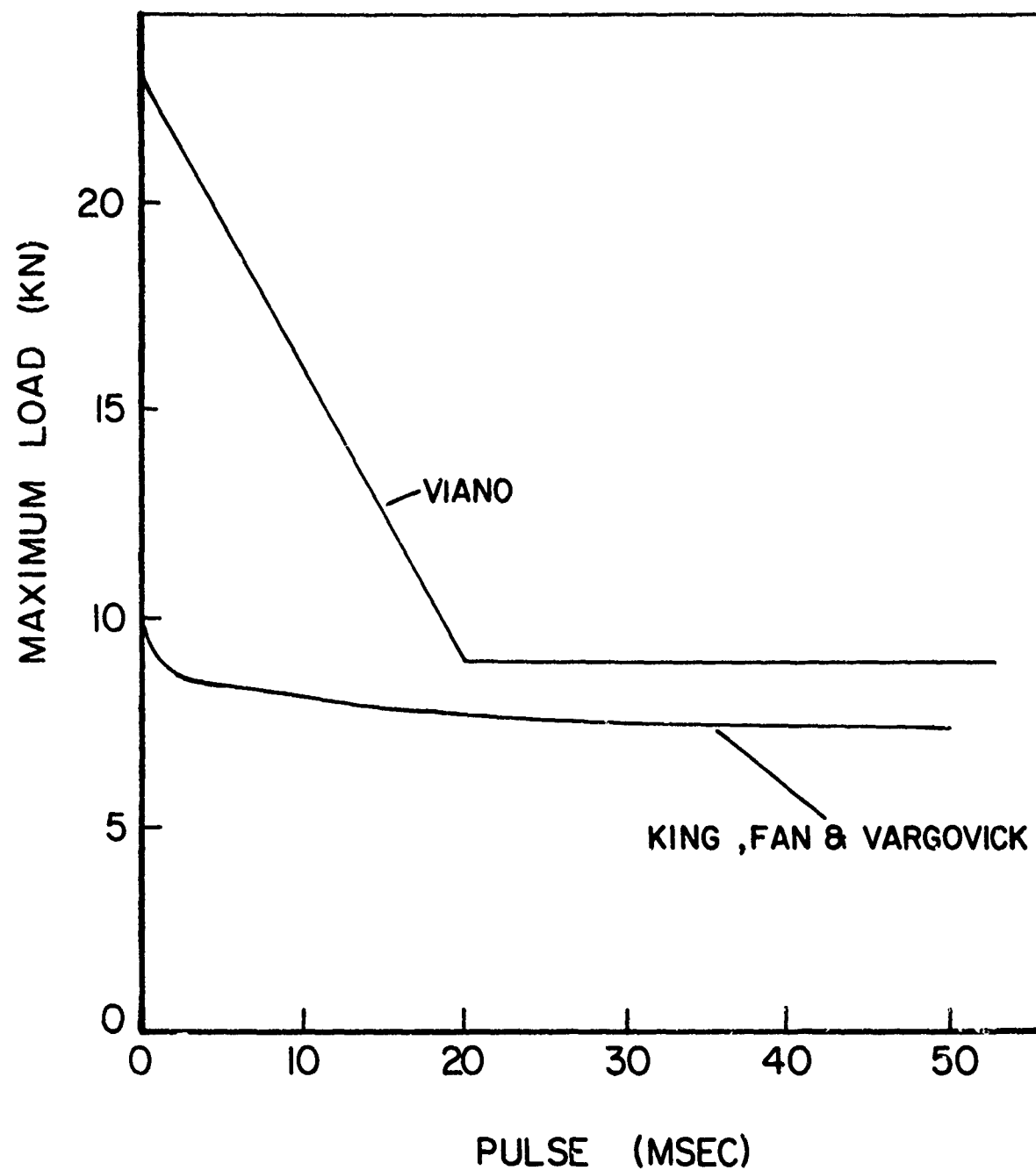
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Figure 1. Time-dependent "Femur Injury Criteria" based on knee impacts.

criterion, etc. (13). Lewis and Goldsmith (14) examined four failure
criteria for compression tests on bone samples, comparing different
loading profiles. The failure criteria examined were: strain at
failure; total work; irreversible work; and cumulative damage index.
None of these criteria were found to be consistent for all load pro-
files, but strain rate effects were not accounted for. A reasonable
correlation does exist between ultimate stress and strain rate when
this is extracted from their data. Since this confirms the previous
work of McElhaney, ultimate stress will be the basis of our criteria.

Despite the fact that the study of material properties of bones
has received considerable attention in the literature throughout the
last 30 years, there is still very little in the way of universal
agreement. A look at a few of the many review papers and books on
the subject shows an amazing array of results (15, 16, 17, 18). Part
of the problem lies in the large number of variables involved. For
example; source of bone - human, canine, bovine, etc.; condition of
bone - dry, wet, embalmed, fresh; subject variations - height, weight,
health, sex, age, etc; whole bone vs bone sample; and so on.

Table 1 gives some idea of the variations presented in the
literature for the restrictive case of fresh human femur bone. Data
from strain-rate-dependent tests is much less available and more
variable.

For this project, we would ideally like data from tests on fresh
human bone samples, tested to failure over a wide range of strain
rates (~6 orders of magnitude), in tension, compression and shear.
Unfortunately, such data does not exist, and the closest approximation
comes from the previously mentioned work by McElhaney. This is a
classic study, often cited, but it will be discussed in some detail
because it is used as the basis for the fracture criteria of this
project.

Figure 2 shows a reproduction of the stress strain curves ob-
tained by McElhaney for various strain rates in compression. This is
for embalmed human compact bone. Ultimate stress, ultimate strain
and elastic modulus are clearly dependent on the rate of loading. A
reasonable approximation to the relation between ultimate stress and
strain rate can be found through a semi log plot. This is the
already-mentioned equation 1

$$\sigma = 4200 \log T + 33000$$

This is based on data from strain rates between 10^{-3} and 10^{3} sec^{-1}.
Equation 1 is plotted in Figure 3 along with other available time-
dependent data. The figure again demonstrates the scatter of data
from various investigators.

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TABLE 1

FRESH HUMAN COMPACT BONE FROM FEMURS, TESTED STATICALLY AND STRESSED IN THE LONGITUDINAL DIRECTION

$$(\times 10^6 \text{ N/m}^2 / \times 10^3 \text{ psi})$$

Tension	Compression	Shear	Bending
122/17.7*	159/23.1*	53.1/7.7*	152/22.0[&]
86.5/12.5*	193/28.0[$]	82.4/11.9*	153/22.1[¢]
133/19.3[$]	134.5/19.5[+]	71.6/10.4[$]	157/22.8*
76.2/11.0[+]	210.9/30.6[+]		164/23.8*
78.9/11.4[+]			181/26.2*

Compiled from various sources reported in (*) Reilly and Burstein (18), (+) Evans (15), ($) Reilly and Burstein (19), (&) Mather (20) and (¢) Vose and Kubala (21).

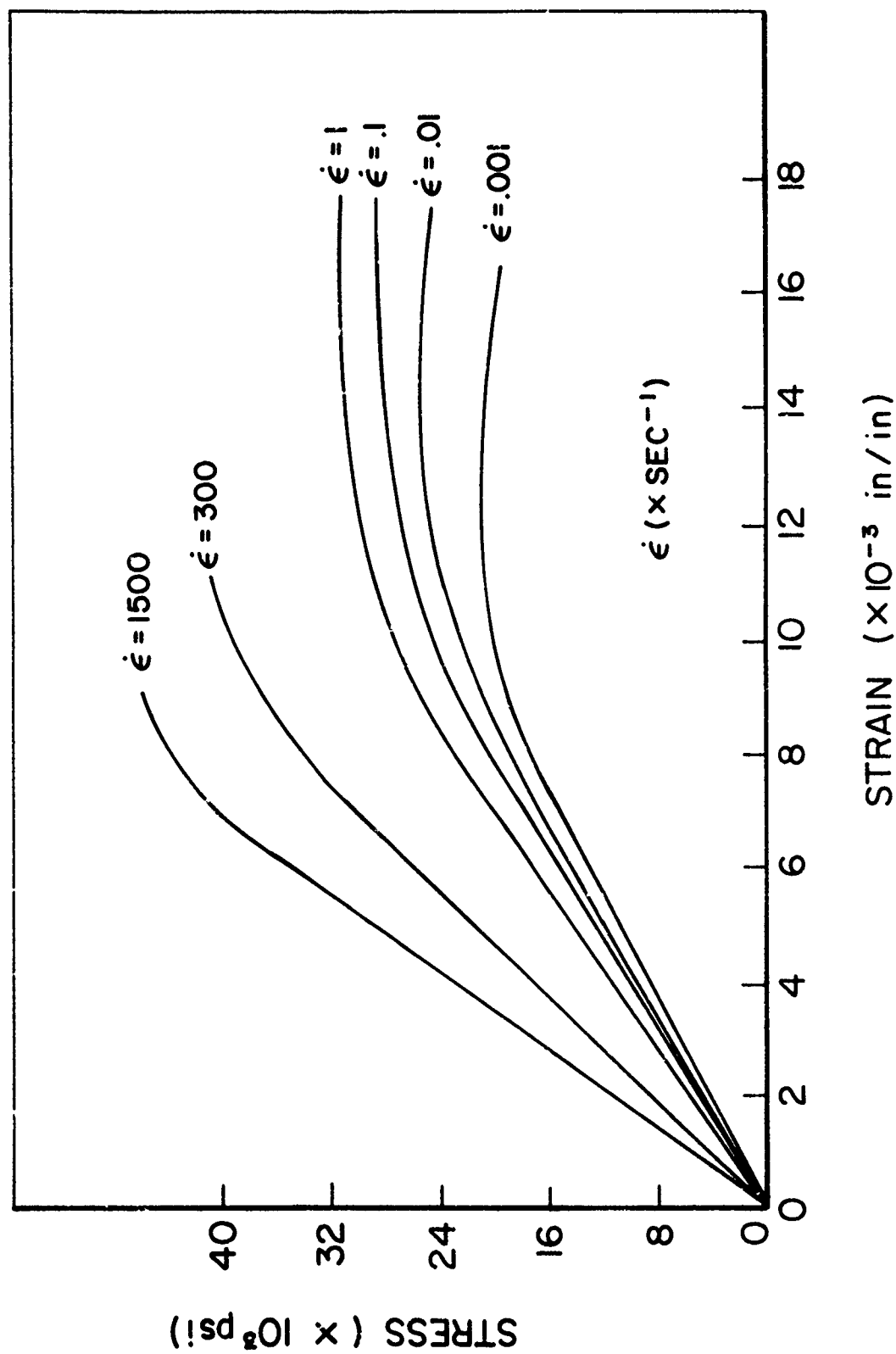Figure 2. Strain rate dependent stress-strain curves for compression of embalmed human compact bone (McElhaney (11)).

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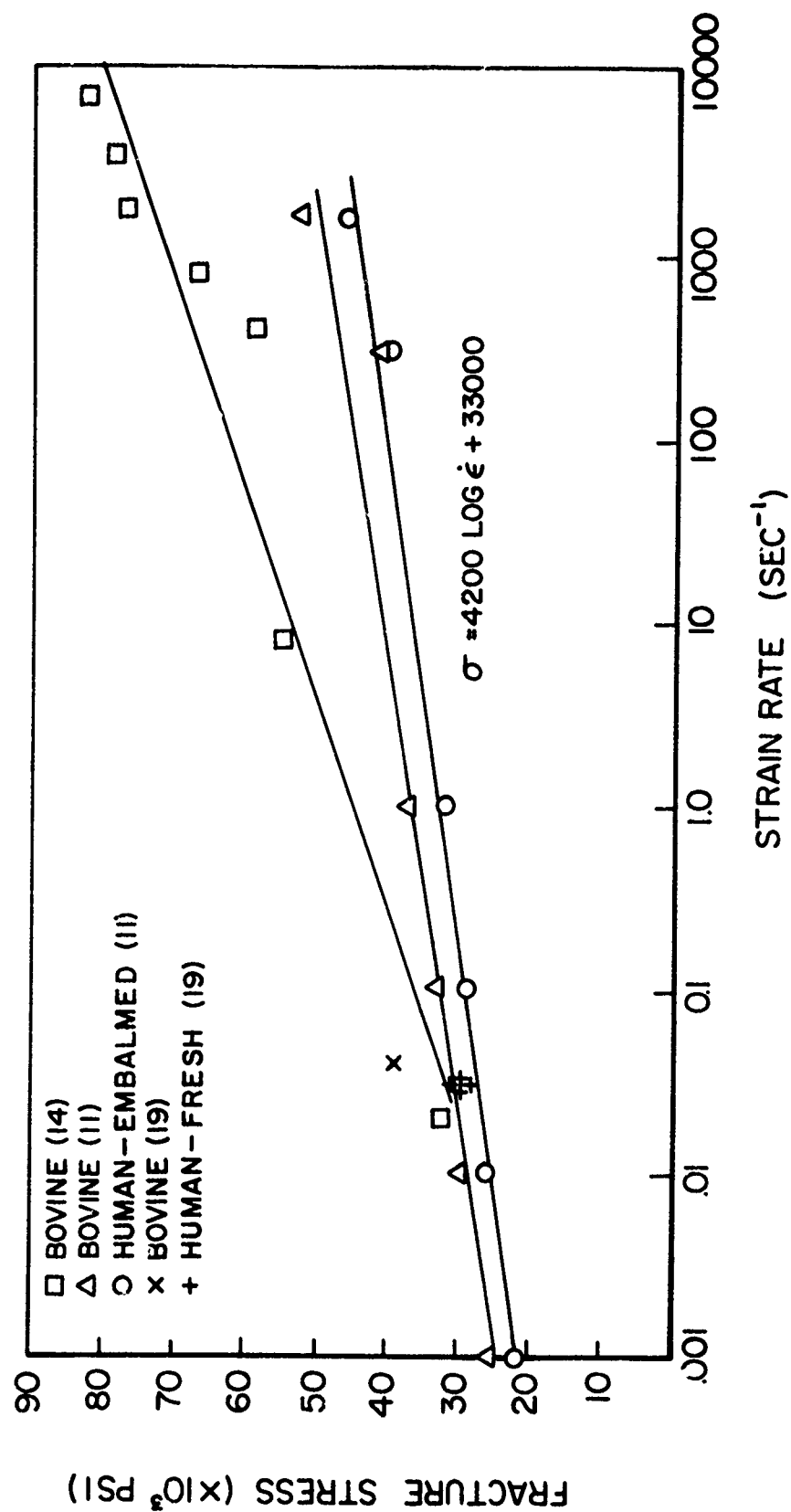Figure 3. Semilog plot of fracture stress vs. strain rate from various authors.

$\sigma = 4200 \ LOG \ \dot{\epsilon} + 33000$

FRACTURE STRESS ($\times 10^3$ PSI)

STRAIN RATE (SEC^{-1})

☐ BOVINE (14)
△ BOVINE (11)
○ HUMAN—EMBALMED (11)
✕ BOVINE (19)
+ HUMAN—FRESH (19)

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Compression tests are the simplest to perform on small samples because this mode eliminates problems of grabbing the specimen. Largely due to these fixation problems, there is no available data on time-dependent ultimate properties of samples in tension or shear over a similar range of strain rates.

FRACTURE CRITERIA MODELS

Based on the available information, the decision was made to use ultimate stress as the fracture/injury parameter, and to use the relation between stress and strain rate from equation 1. However, equation 1 could not be directly used with the ATB model.

As mentioned earlier, the ATB model finds motion and joint and contact forces for the limbs. This information is sufficient to calculate the stresses in the bones if the geometry is known. For this project, owing to time and information constraints, a very simplified geometry was assumed. Each long bone (femur, tibia, humerus, ulna) was treated as if it were a straight, hollow cylinder with uniform, isotropic properties. The geometric properties (length, cross-sectional area, moment of inertia and outer radius) were approximated to be consistent with the dimensions of the subject in the simulation (in this case a 95th percentile man).

To compare the calculated stresses to the allowable stress (equation 1), the strain rate is needed. In order to determine the strain rate from the available information, one would need to know the stress-strain relations for all rates. This is not directly available, so several approximations are needed to obtain a relation between stress and strain rate.

Going back to the McElhaney data, we begin by looking for a relation between the apparent elastic modulus (based on the linear portion of the stress-strain curve) and the strain rate. A fit of the data on a log-log plot yields the following approximate relation

$$\text{Log } E = 0.067 \text{ Log } \dot{\epsilon} + 6.52$$

or
(4)
$$E = 3.311 \times 10^6 \; \dot{\epsilon}^{0.067}$$

If we assume a linear stress-strain relation

$$\sigma = E\epsilon$$
(5)

(this is clearly no longer true in the plastic region), then equations 4 and 5 can be combined to yield

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$$\sigma = 3.311 \times 10^6 \; \dot{\epsilon}^{0.067} \; \epsilon \tag{6}$$

which can be differentiated with respect to time to give the stress rate

$$\dot{\sigma} = 3.311 \times 10^6 \left\{ \dot{\epsilon}^{0.067} \; \dot{\epsilon} + \epsilon(0.067 \; \dot{\epsilon}^{-.933} \; \ddot{\epsilon}) \right\} \tag{7}$$

Since the tests were done at constant strain rate, $\dot{\sigma}$ becomes

$$\dot{\sigma} = 3.311 \times 10^6 \; \dot{\epsilon}^{1.067}$$

which can be inverted to form

$$\dot{\epsilon} = 7.752 \times 10^{-7} \; \dot{\sigma}^{0.937}$$

or

$$\log \dot{\epsilon} = 0.937 \log \dot{\sigma} - 6.111 \tag{8}$$

Combining equations 8 and 1 gives the desired relation between ultimate stress and strain rate

$$\sigma = 3936 \log \dot{\sigma} + 7336 \tag{9}$$

Since the stress is known as a function of time, we can approximate the stress rate simply by

$$\dot{\sigma} = \Delta\sigma/\Delta t, \tag{10}$$

where Δt = current ATBM solution time increment
and $\Delta\sigma$ = change in stress during Δt.

Equations 9 and 10 give us a relationship for the allowable stress in terms of the available parameters.

If $\dot{\epsilon}$ = .001 sec^{-1} is taken as a "static" strain rate, the corresponding static stress rate is $\dot{\sigma}$ = 2.084 x 10^3 psi/sec from equation 7. Both equations 1 and 9 predict an allowable stress of 20,400 psi for this stress rate. This value is below the average compressive stress reported for fresh human bone of 25,000 psi (see Table 1). Recall that the McElhaney data is from embalmed subjects.

If it is assumed, for lack of any real data, that fresh bone would behave in much the same was as embalmed bone, a simple offset correction can be made to equation 9 to bring it in line with available fresh bone data. The same argument can also be made for tension and shear stresses, leading to the following adjusted relations

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Compression $\qquad \sigma_c = 3936 \log \dot{\sigma}_c + 12000$

Tension $\qquad \sigma_T = 3936 \log \dot{\sigma}_T + 1407 \qquad\qquad$ (11)

Shear $\qquad \tau = 3936 \log \dot{\tau} - 2993$

Equations 11 define the rate-dependent allowable stresses to be used as the fracture criteria in the program.

A similar approach, which is more in line with the reported car crash criteria, uses stress pulse duration rather than stress rate. This can be approximated by assuming a sinusoidal wave form.

Thus

$$\sigma = A(1-\cos \omega t) \qquad\qquad (12)$$

and

$$\dot{\sigma} = A\omega \sin \omega t \qquad\qquad (13)$$

and, in terms of pulse length,

$$\omega = \frac{2\pi}{T} \qquad\qquad (14)$$

Using the maximum values for equations 12 and 13 along with 14 in, for example, the compression part of equation 11 leads to

$$\sigma_{max} = 3936 \log \dot{\sigma}_{max} + 12000$$

hence,

$$2A = 3936 \log \left(\frac{2\pi A}{T}\right) + 12000 \qquad\qquad (15)$$

Equation 15, and similar expressions for tension and shear, can be solved numerically for 2A, the amplitude of allowable stress, given a pulse length T. This provides an alternate measure of allowable stress.

PROGRAM DESCRIPTION

The current version of the program, BREAK, has been designed to run separately from the ATB program. A disk file was generated using existing options in the ATB program which includes, at 10 msec intervals, the linear and angular displacement, velocity and acceleration histories of all segments, and joint forces and torques at each joint, and all contact forces from allowable contact situations. This file, in fact, contains much more information than is needed, and much of what is desired is with respect to inconvenient coordinate systems. This will be fairly easily remedied once the utility of BREAK has been improved and the format is settled. However, a significant portion of the code in BREAK is involved with overcoming the current format problems. The subroutines which search the file for appropriate forces and put them into the desired coordinate system are largely superficial to the problem and hence will not be discussed.

The remaining important functions of the program involve the calculation of stresses in the bones, the calculation of allowable stresses, and the output of results. As mentioned earlier, the bones are considered to be uniform, hollow cylinders, and thus the stress calculations are straightforward. Only axial tension and compression (due to axial and bending loads) and shear due to torsion are considered. The following equations define the stresses at a particular cross section ($x = x_p$),

$$\sigma_x = \frac{r_x}{A} + \Sigma \frac{CF_{x_i}}{A} + \frac{r}{I} \left\{ \left[M_z - F_y X_p - \Sigma (CF_{y_i} (X_p - X_i) \langle X_p - X_i \rangle \right. \right.$$

$$\left. + CF_{x_i} Y_i \langle X_p - X_i \rangle) \right]^2 + \left[M_v + F_z X_p + \Sigma (CF_{z_i} (X_p - X_i) \langle X_p - X_i \rangle \right.$$

$$\left. \left. + CF_{x_i} Z_i \langle X_p - X_i \rangle) \right]^2 \right\}^{1/2} \tag{16}$$

$$\tau = \left\{ M_x + \Sigma (CF_{z_i} Y_i - CF_{y_i} Z_i) \langle X_p - X_i \rangle \right\} \frac{r}{J} \tag{17}$$

where F_x, F_y, F_z, M_x, M_y, M_z are joint forces and moments

CF_{x_i}, CF_{y_i}, CF_{z_i} are contact forces due to ith contact

X_i, Y_i, Z_i are coordinates of the ith contact

$\langle X_p - X_i \rangle = 0$ for $X_p < X_i$; $= 1$ for $X_p \geq X_i$

r, I, J are the radius and moments of inertia.

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The axial stresses are found for diametrically opposite points and the principal stresses determined so that the maximum tensile, compressive and shear stresses can be obtained. The same stress calculations are carried out at some number of points along the bone (the number being a program option) at each time step and only the maxima are recorded. The stress-time history which is produced is therefore not the stress at any particular point, but the maximum stress anywhere along the segment at each time. This may present some inconsistencies for localized forces and will have some tendency to reduce stress rate effects. These items should be examined further.

The calculation of allowable stresses involves a straightforward application of equation (11) for the stress rate test. For the pulse test, two additional features are needed. First, pulse length is estimated by monitoring two changes in sign of the stress rate, or the time between two "valleys" in the stress. The pulse corresponding to the maximum stress only is evaluated. The second additional feature involves the solution of equation 15, which is carried out using a standard Newton Raphson root finding routine.

The output and plotting routines are standard and need not be discussed.

SIMULATION RESULTS

A very limited series of sample runs has been made to demonstrate the program, BREAK. As mentioned earlier, an existing simulation using the ATB model was placed on a computer file which could be accessed by BREAK. The simulation used was a 95th percentile male, ejected from a high-speed aircraft and subjected to high acceleration loading and high wind forces. The first 200 msec of this event were examined. For this report the injury analysis results for the left lower arm (LLA) will be presented. This particular segment was chosen because of the occurrence of high joint forces and a contact pulse.

Figures 4 and 5 show the joint forces and moments, respectively, from the proximal joint of the segment. Note some fairly high frequency components to the loading. Contact forces are shown in Figure 6. No attempt is made in this figure to indicate where along the axes contact is taking place. The impact can be seen at 0.07 seconds. Figures 7, 8 and 9 compare the highest stress at each time step with the corresponding calculated allowable stress for compression, tension and shear. Note that the allowable stress is a definite function of stress rate, and that there is a beneficial raising of the allowable stress for high-stress spikes. The allowable stresses predicted by the pulse technique are also shown in these figures. The pulse technique is slightly less conservative and is sensitive to the definition of the pulse length.
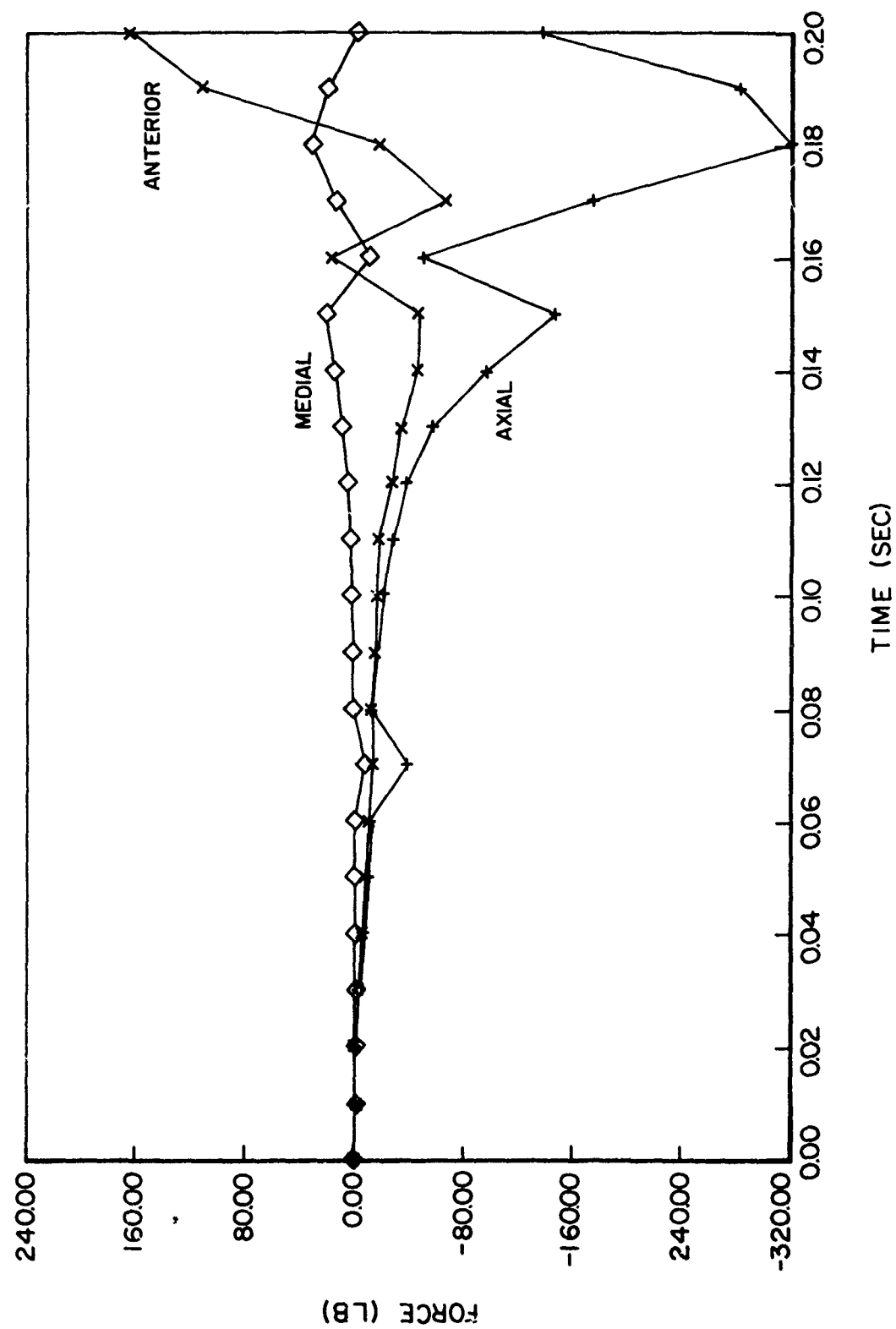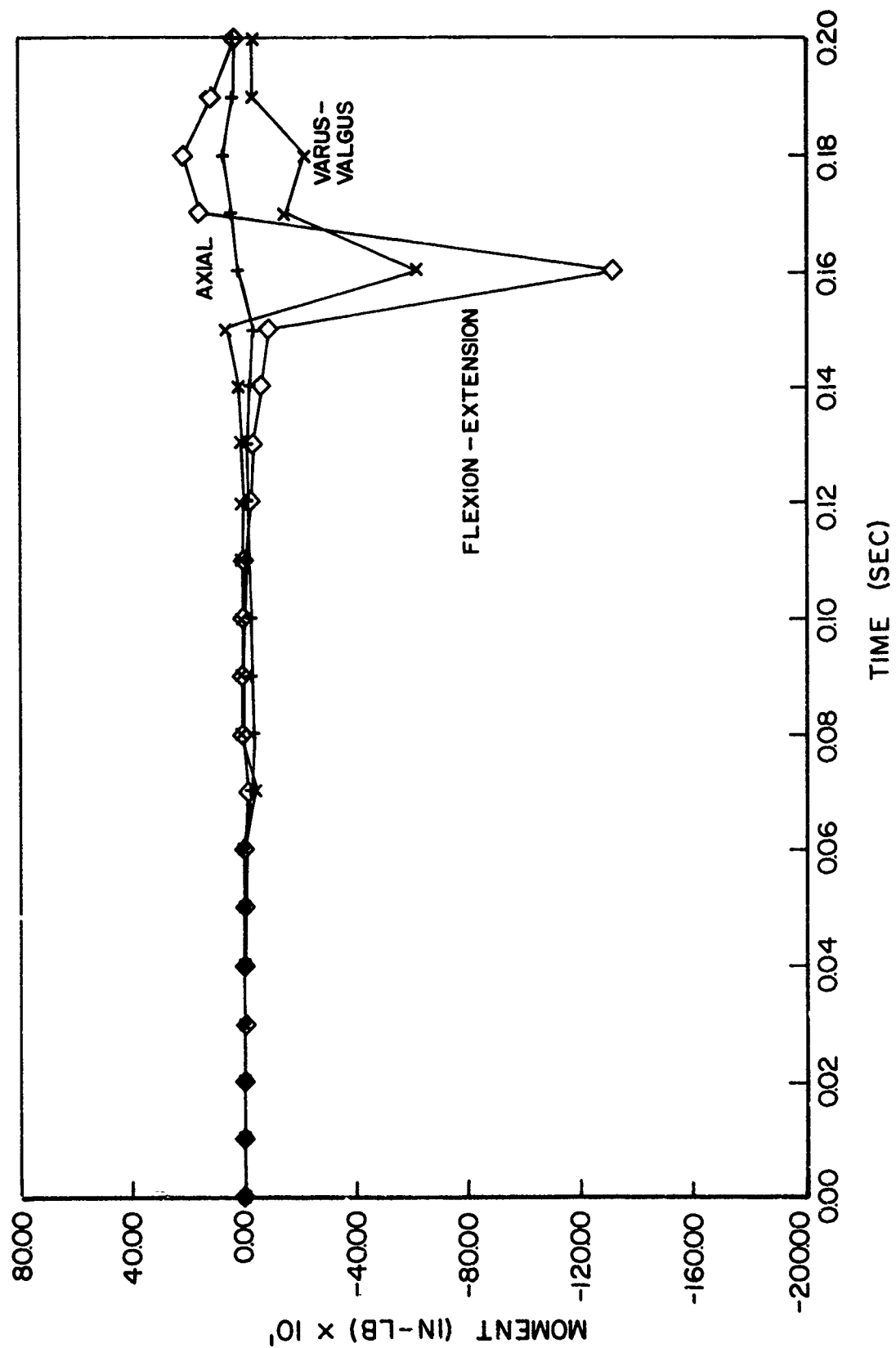
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Figure 4. Joint forces.
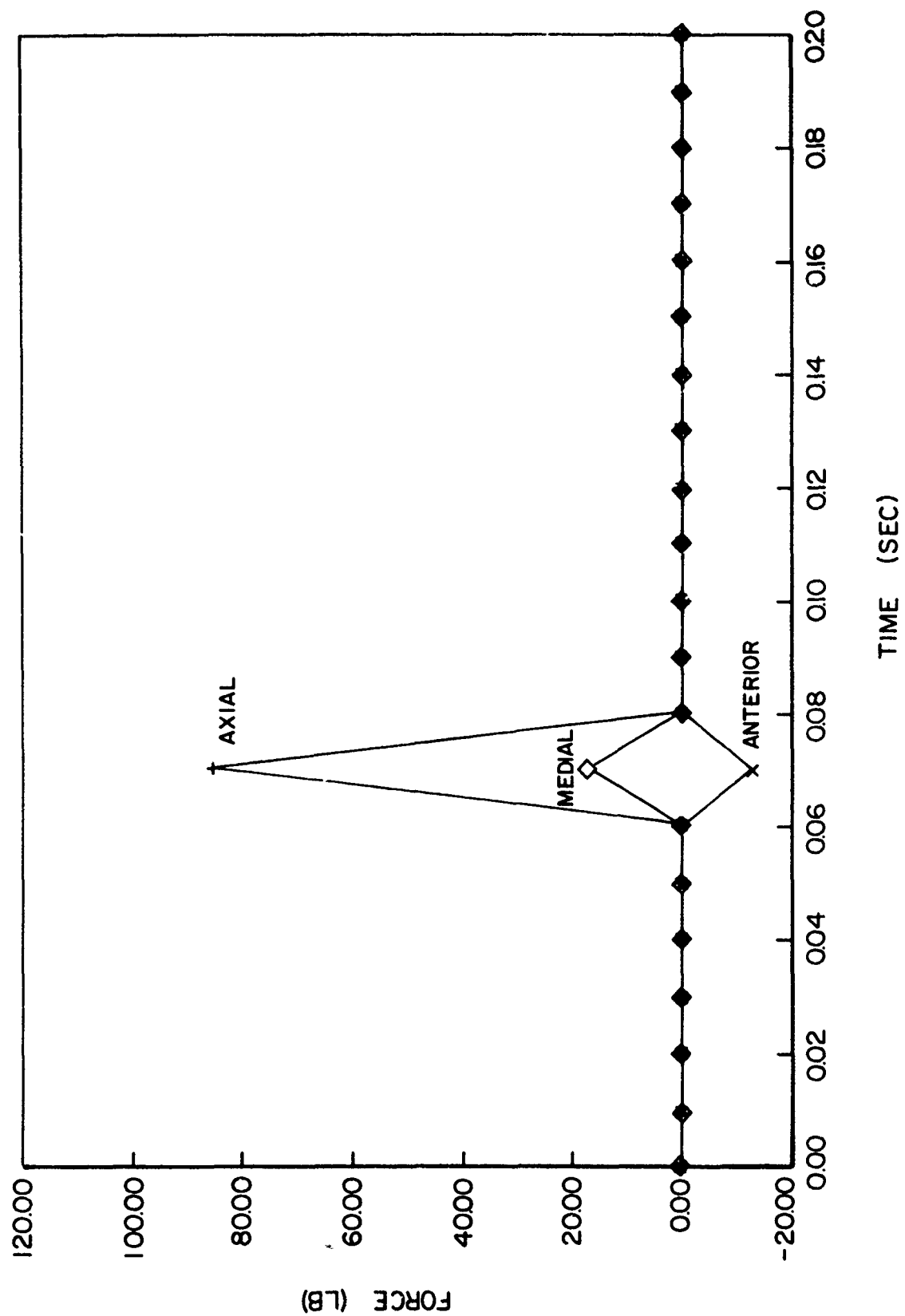
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Figure 5. Joint moments.
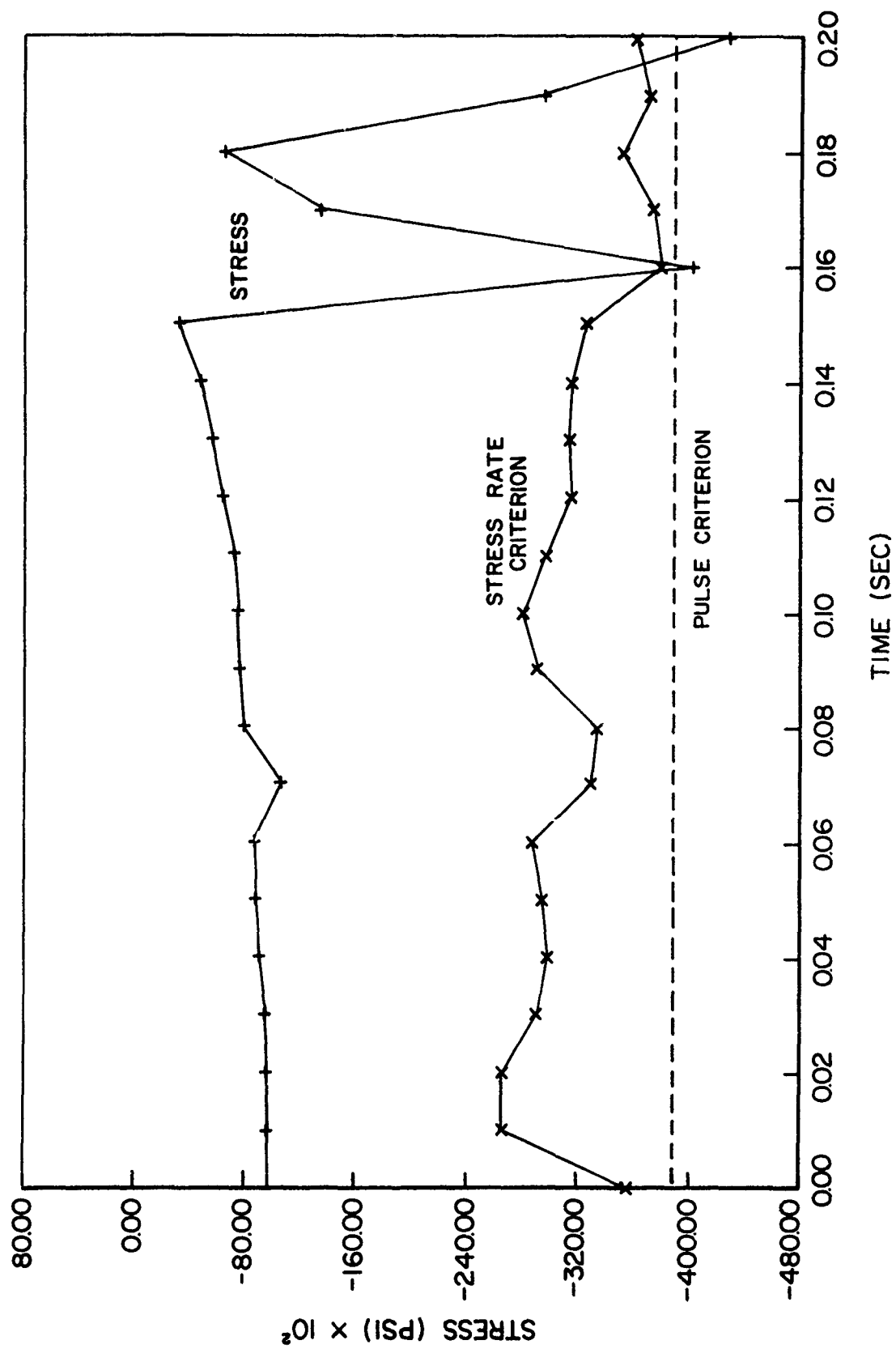
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Figure 6. Contact forces.
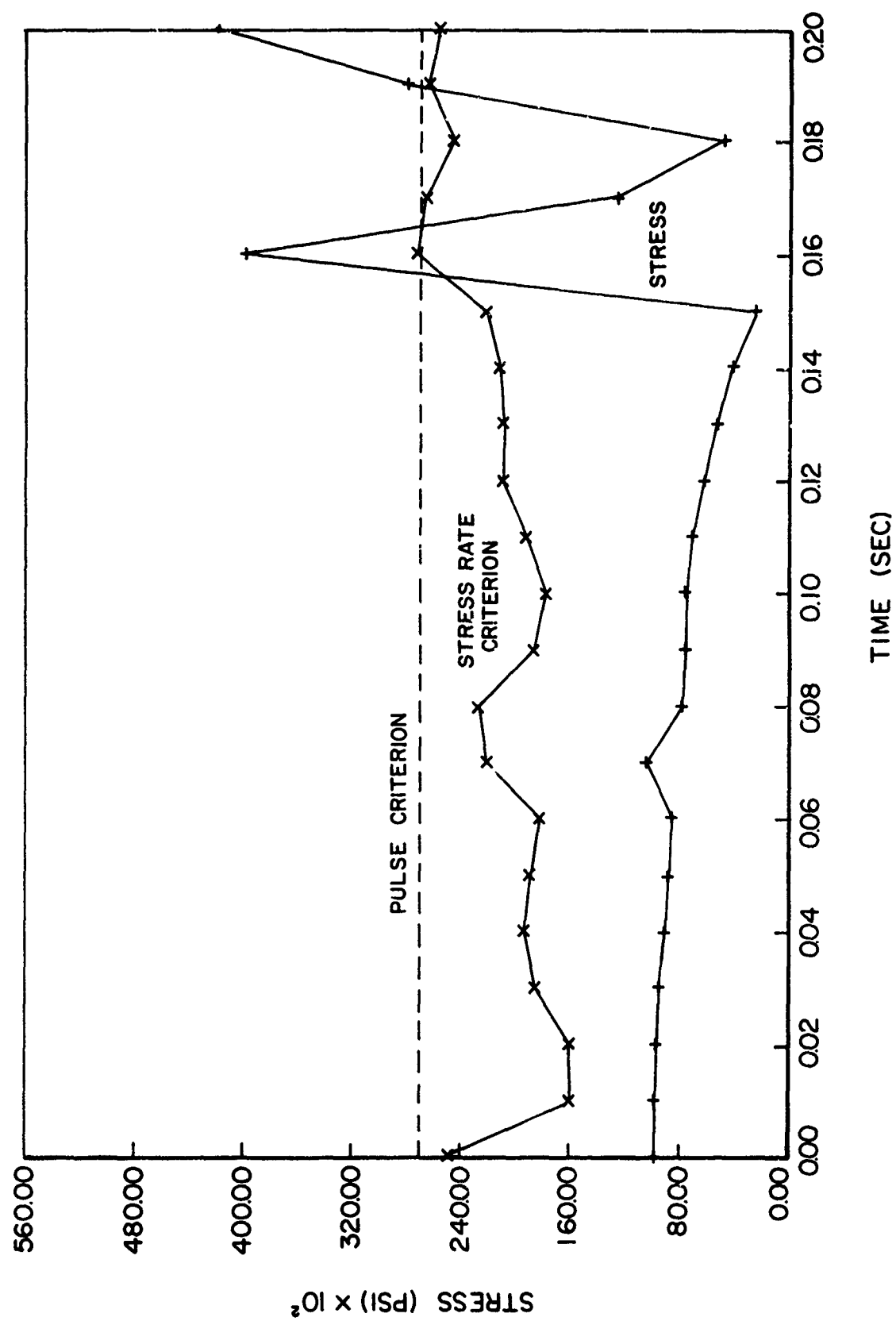
Figure 7. Compressive stress versus allowable stress.

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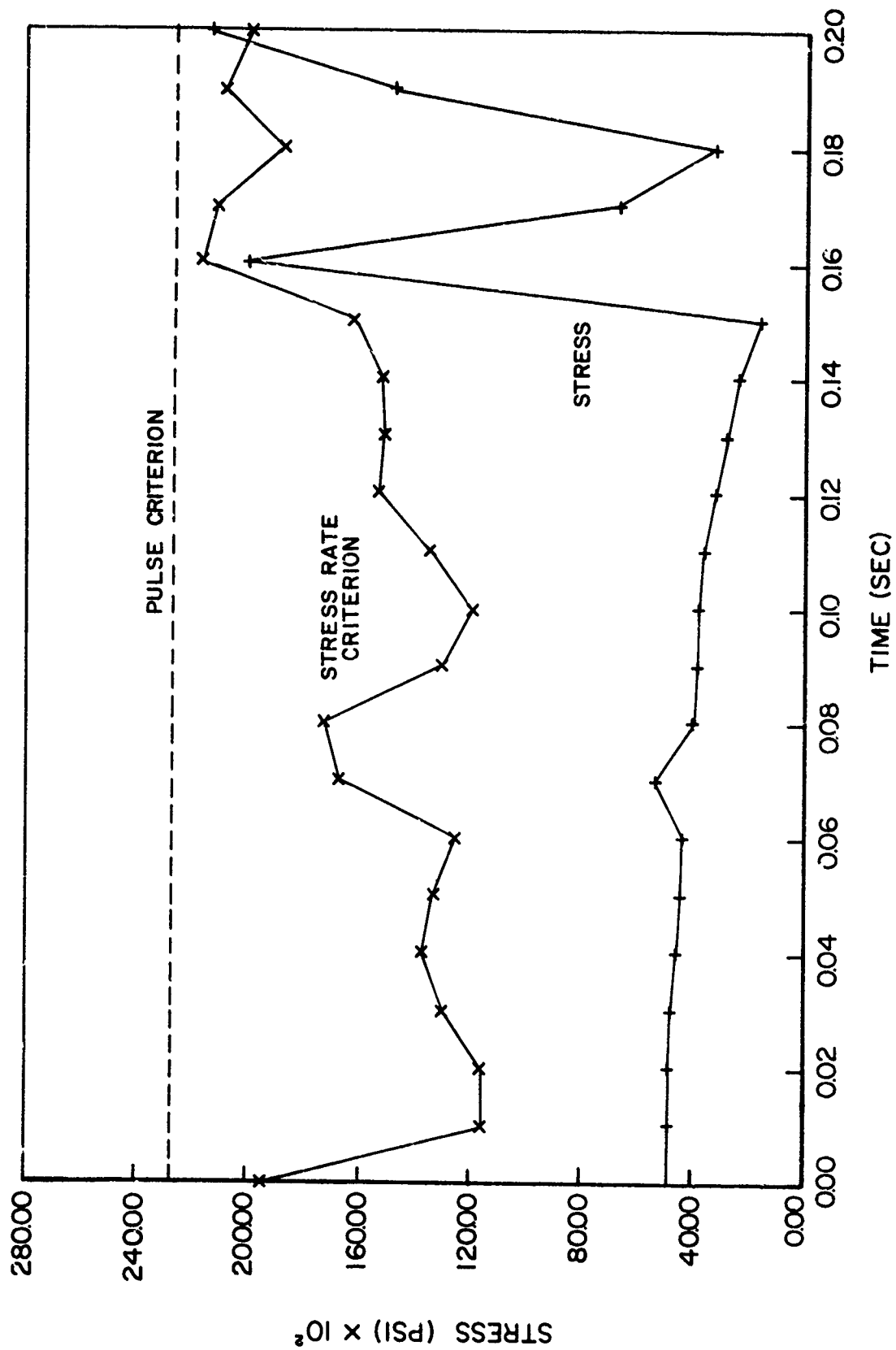Figure 8. Tensile stress versus allowable stress.

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Figure 9. Shear stress versus allowable stress.

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This example demonstrates the capability of the program to evaluate the time-varying failure criteria within the context of the ATB model. The validity of the failure criteria have not been shown.

RECOMMENDATIONS

The two primary goals of this research effort have been accomplished, that is, to choose appropriate long-bone failure criteria and to implement a program which would evaluate limb loading from the ATB model with respect to those criteria. On the basis of this preliminary effort there are at least two major areas open to further study. These come broadly under the areas of improvement of the current program, and its expansion and verification.

The current program is a demonstration program which has not been fully developed. The most serious shortcoming is the already mentioned difficulty with the output file from the ATB. Informal discussions with one of the authors of the ATB model have taken place concerning setting up a specific file containing only that information needed by BREAK, and with reference to the correct coordinate systems. This can be easily accomplished and will greatly improve the efficiency of the current program. This should be implemented and corresponding adjustments made to BREAK.

There are also numerous minor changes to the program which should be made for clarity, efficiency and versatility, including changing output formats, subroutine reorganization, easier user control of options, etc. Once these improvements have been made, complete documentation should follow. In addition, other simulations should be studied over a wide range of conditions.

The second major area of improvement, extension and verification, involves both going back to the original choice of the fracture criteria and examining, in detail, their implications, and going beyond the scope of the current model to look at such things as joint injury and the impact of statistical variations in the data. Throughout the development of these criteria, simplifying assumptions were made without, necessarily, being fully justified. (For example, simplified bone geometry eliminates axial-bending coupling in the femur due to axis curvature.) These assumptions should be checked thoroughly and altered as needed to provide more comprehensive and accurate criteria. Ideally, this would be done in conjunction with a testing program, particularly with regard to basic material properties, but in the least it would attempt to accurately model existing data. On a grosser scale, the two proposed Femur Injury Criteria could be compared to results using BREAK through car crash simulations with the ATB model. The questions of joint injury and the statistical variations in all the data and their impacts on the injury criteria have not as yet been considered.

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